Characterization of the nanoDot OSLD dosimeter in CT

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(Received 17 February 2014; revised 6 February 2015; accepted for publication 12 February 2015; published 20 March 2015)

Purpose: The extensive use of computed tomography (CT) in diagnostic procedures is accompanied by a growing need for more accurate and patient-specific dosimetry techniques. Optically stimulated luminescent dosimeters (OSLDs) offer a potential solution for patient-specific CT point-based surface dosimetry by measuring air kerma. The purpose of this work was to characterize the OSLD nanoDot for CT dosimetry, quantifying necessary correction factors, and evaluating the uncertainty of these factors.

Methods: A characterization of the Landauer OSL nanoDot (Landauer, Inc., Greenwood, IL) was conducted using both measurements and theoretical approaches in a CT environment. The effects of signal depletion, signal fading, dose linearity, and angular dependence were characterized through direct measurement for CT energies (80–140 kV) and delivered doses ranging from ~5 to >1000 mGy. Energy dependence as a function of scan parameters was evaluated using two independent approaches: direct measurement and a theoretical approach based on Burlin cavity theory and Monte Carlo simulated spectra. This beam-quality dependence was evaluated for a range of CT scanning parameters.

Results: Correction factors for the dosimeter response in terms of signal fading, dose linearity, and angular dependence were found to be small for most measurement conditions (<3%). The relative uncertainty was determined for each factor and reported at the two-sigma level. Differences in irradiation geometry (rotational versus static) resulted in a difference in dosimeter signal of 3% on average. Beam quality varied with scan parameters and necessitated the largest correction factor, ranging from 0.80 to 1.15 relative to a calibration performed in air using a 120 kV beam. Good agreement was found between the theoretical and measurement approaches.

Conclusions: Correction factors for the measurement of air kerma were generally small for CT dosimetry, although angular effects, and particularly effects due to changes in beam quality, could be more substantial. In particular, it would likely be necessary to account for variations in CT scan parameters and measurement location when performing CT dosimetry using OSLD. © 2015 American Association of Physicists in Medicine. [<http://dx.doi.org/10.1118/1.4914398>]

Key words: OSLD, dosimetry, CT, imaging, point dose

1. INTRODUCTION

Imaging with computed tomography (CT) is performed with ever-increasing frequency, with up to 80 million scans being performed each year in the US alone.¹ Despite the numerous benefits offered by this modality, CT studies are not without risk; radiation exposure carries the risk of both acute and long-term effects for sensitive organs, especially among children where the risk can be three times that for adults.^{2–7}

Optically stimulated luminescent dosimeters (OSLDs) offer a potential method for fast and accurate point-based patient-specific CT dosimetry at the patient's surface. These integrating dosimeters store dose information during irradiation (electrons in the crystal structure of the dosimeter are elevated to metastable energy traps). At dosimeter readout, the crystal is stimulated with a light-emitting diode (LED), allowing the electrons to fall back to their original energy state while emitting characteristic light proportional to the amount of absorbed radiation dose. One type of Al_2O_3 -based OSLD, the nanoDot (Landauer, Inc., Glenwood, IL), is commercially available and is small, robust, reusable, has high sensitivity, and has no impact on image quality,⁸ making it a realistic choice for point measurements in diagnostic imaging.

While Al_2O_3 based OSLDs have been characterized in radiotherapy environments,^{9–16} the substantially lower doses and photon energy used in CT imaging result in different characteristics of this dosimeter. Recent publications have provided some data on the use of OSLDs in diagnostic settings,^{17,18} including cone-beam CT scans associated with image guidance for therapeutic procedures.^{19,20} These initial works have demonstrated that this dosimeter has potential as a clinical dosimeter in a diagnostic CT environment, and groups have begun using this detector for such purposes.^{21,22} However, no work, to the authors' knowledge, has systematically evaluated all of the correction factors potentially necessary for accurate dosimetry in a CT environment.

One major concern is potential angular dependence of the OSLD nanoDot.^{17,18,23} Al-Senan and Hatab reported angular effects of up to 10% in CT applications.¹⁷ While these investigations have highlighted the potential need to apply correction for angular dependence, the dependence on irradiation angle in rotating beams, particularly as it relates to the calibration conditions and clinical use of the dosimeters, requires further investigation.

A second major concern is that Al_2O_3 over-responds to low energy photons (with respect to water or tissue) because of its relatively large effective atomic number.^{10,14,16,17,24} Per milligray of dose delivered to water, the response of Al_2O_3 can be larger by a factor of 3.5 or more due to increased photoelectric effect at low energies. While this issue has been identified in the literature,^{24–26} and has been examined for some cone-beam CT scans,²⁰ a thorough study of appropriate correction factors in a diagnostic CT environment has not been done. Of particular interest is the variation in energy response as a function of scan parameters or measurement position. Such factors (including kV, position on phantom, size of phantom, and scan extent) may have a significant influence on the dosimeter response^{20,24} and should be accounted for.

Therefore, further characterization of this detector is needed to ensure its accuracy in CT dosimetry. Because of the relatively limited experience with this dosimeter in a CT environment, we conducted a full characterization of the OSLD nanoDot, examining signal depletion, signal fading, and dose linearity as well as angular dependence and beam-quality dependence. We also quantified the inter- and intradosimeter reproducibility of this dosimetry system in a CT environment.

2. METHODS AND MATERIALS

All CT irradiations were performed on a 64-slice CT scanner (Discovery CT750 HD, GE Healthcare; Milwaukee, WI). The OSLDs evaluated in this study were Landauer nanoDots ($\text{Al}_2\text{O}_3\text{:C}$), which are thin discs of diameter 4 mm and thickness 0.3 mm. The active dosimeter volume is enclosed in a plastic cassette measuring $10 \times 10 \times 2$ mm (Fig. 1).

All dosimeters were read using a single InLight MicroStar OSL reader (Landauer, Inc., Glenwood, IL) which was operated in continuous wave mode for a 7 s read time. The MicroStar reader has two readout modes: a strong beam (low doses) or weak beam (high doses). For the purposes of this study, the strong LED beam (low dose) was used for all readings except as described for the linearity test.

The MicroStar reader was calibrated using dosimeters irradiated to a known dose in air using the 120 kV CT beam. Calibration OSLDs (i.e., standards) were irradiated for each session of experimental measurements, such that each reading session had a unique system calibration factor (N_D). The known dose given to these OSLDs was measured by irradiating a 0.6 cm^3 farmer-type ion chamber (RadCal Model 10 $\times$ 5) with a NIST-traceable calibration coefficient (using identical irradiation conditions). The chamber reading (q) was scaled by the calibration coefficient (N_k) and was also corrected for temperature and pressure effects (P_{TP}) and the electrometer factor (P_{elec}), per AAPM Task Group 111 [Eq. (1)].²⁷ Dose to air (D_{air}) and air kerma (K_{air}) are approximately equal because of the local deposition of charged secondary particles and are solved for as follows:

$$D_{\text{air}} \approx K_{\text{air}} = q P_{TP} P_{\text{elec}} N_k. \quad (1)$$

OSLD dosimetry is based on the raw signal and a series of correction factors. First, as per Eq. (2), the raw OSLD signal (M_{raw}) must be corrected for signal depletion (k_d) for multiple sequential readouts of the dosimeter ($j = 3$ in this study), and

FIG. 1. OSLD nanoDot: (a) active volume visible when open; (b) front and back of closed dosimeter.

individual element sensitivity ($k_{s,i}$), which is relative to the average sensitivity of the batch. These provide an average corrected dosimeter signal ($\bar{M}_{\text{corr}}$),

$$\bar{M}_{\text{corr}} = \frac{\sum_j (M_{\text{raw},j} k_{d,j})}{j} \times k_{s,i}. \quad (2)$$

Based on experience in a therapeutic environment, the OSLD signal may also need to be corrected for dose nonlinearity (k_L), fading (k_F), irradiation geometry, (k_G), dosimeter angle

(k_θ), and beam quality (k_Q). Therefore, including the system calibration factor (N_D), the dose (D) from this dosimeter is calculated as

$$D = \bar{M}_{\text{corr}} \times N_D \times k_L \times k_F \times k_G \times k_\theta \times k_Q. \quad (3)$$

The calibration factor is determined from the calibration dose measured with the ion chamber [Eq. (1)], giving dose to air in this case, and the corrected signal from the calibration OSLD,

$$N_D = \frac{D_{\text{calibration}}}{\bar{M}_{\text{corr,calibration}} \times k_{L,\text{calibration}} \times k_{F,\text{calibration}} \times k_{G,\text{calibration}} \times k_{\theta,\text{calibration}}}. \quad (4)$$

Characterization of the nanoDot [i.e., determining the correction factors in Eqs. (2) and (3)] was, therefore, performed.

2.A. Basic characterization of the nanoDot

2.A.1. Signal depletion ($k_{d,j}$)

The nanoDot signal is partially depleted during the readout process, which is relevant as the dosimeter is typically read out multiple times sequentially. The amount of depletion depends on the intensity of the LED during the readout process and therefore varies between readers. This signal depletion was determined for 5 nanoDots in the high-intensity (low dose) mode of our reader. Each dosimeter was read 20 sequential times. Signal depletion was measured as the percentage of initial signal lost for each successive reading, and the average depletion as a function of reading number was determined.

2.A.2. Reproducibility

Twenty randomly selected dosimeters were identically irradiated to a dose of ~5 mGy in the 120 kV CT beam, two at a time, in a custom-built solid holder that snugly fit the OSLD (Fig. 2). This holder was placed in the center chamber of an acrylic pediatric head CTDI phantom (10 cm diameter by 15 cm long). To test a range of doses, these dosimeters were optically bleached and then this process was repeated, irradiating these dosimeters to ~50 mGy. Because of much greater familiarity in the literature with this dosimeter's behavior in cobalt beams, these dosimeters were also irradiated to ~90 and ~1000 mGy in a ^{60}Co beamline to compare the reproducibility between the different beam energies and dose levels.

Each individual dosimeter was read out three successive times, and each reading was corrected for signal depletion and element sensitivity ($k_{s,i}$). The interdot reproducibility of the system (dosimeter and reader) was expressed as the coefficient of variation of the signal across the set of 20 dosimeters. Over the 20 dosimeters evaluated, the maximum intradot variability (largest coefficient of variation over the three readings of each dosimeter) was also recorded.

2.A.3. Signal fading (k_F)

The signal stored in the dosimeter fades with time as trapped electrons, even at room temperature, can spontaneously escape from the energy traps.¹² This signal fade is particularly large immediately after irradiation. In megavoltage beams, fading is substantial within the first 10 min following irradiation but is then minimal.^{10,26} This has not previously been well characterized for irradiations relevant to a diagnostic environment. Therefore, we irradiated 40 nanoDots in a 120 kV CT beam under identical conditions (using the custom acrylic holder shown in Fig. 2 placed in the center chamber of a 10 cm diameter CTDI phantom) to a dose of approximately 50 mGy and read them out at specific time points between 30 min and 18 days postirradiation. Because the reader was located in a different building than the CT scanner, it was not possible to read the dosimeters less than 30 min post irradiation.

FIG. 2. Custom OSLD nanoDot acrylic holder used for CT measurements: (a) open and (b) closed.

2.A.4. Dose-response linearity (k_L)

OSLDs exhibit a linear response at low doses, but a supralinear dose response at higher doses; however, it is unclear at exactly what dose the response becomes nonlinear. Moreover, the linearity characteristics of nanoDots have been found to be dependent on the energy of the beam used to irradiate the dosimeter.²⁸ This is of particular concern in low-energy environments such as CT because the high atomic number of the OSLD means that a given physical dose corresponds to a much greater trapped signal than it would in a high-energy (e.g., therapeutic) environment. Therefore, to determine if there is any supralinearity in the response at doses and energies relevant to CT dosimetry, the linearity of the nanoDot was evaluated between 42 and 1162 mGy (all at 120 kV). The relationship between the dosimeter signal and the delivered dose was evaluated to determine whether a nonlinearity correction was needed.

2.B. Dependence on irradiation geometry (k_G)

Previous studies have suggested that the OSLD nanoDot signal may be dependent on the incident angle of irradiation.^{17,23} Because CT employs a rotating x-ray tube, there may be a loss of signal resulting from the x-ray beam incidence on the sides (edges) of the nanoDot (e.g., as would occur briefly during times in a rotating exposure) relative to a beam that is only perpendicularly incident (toward the “flat” side of the dosimeter, as would occur during a localizer scan or potentially during dosimeter calibration) (illustrated in Fig. 3).

To determine the OSLD response for rotational irradiations versus planar irradiations, OSLDs were irradiated at the same location under each set of conditions (as shown in Fig. 3), and the signal from each was normalized to an ion chamber response under identical conditions. These measurements were performed under three different conditions: in air (at machine isocenter), on top of a phantom (a 16 cm diameter CTDI phantom), and inside a phantom (the center of a 16 cm

diameter CTDI phantom). These three locations were evaluated to investigate the impact of different scatter conditions (from different potential clinical uses and calibrations) on the irradiation geometry response.

2.C. Angular dependence (k_θ)

For clinical CT dosimetry with the nanoDot, the angle of the dosimeter relative to the CT bore may not always be constant. For patient dosimetry measurements, the dosimeter may be placed on a body contour such that the dosimeter will be tilted relative to the CT bore. This is a different issue than in the paragraph above and requires separate evaluation. Most often, the OSLD would be placed such that the active volume is perpendicular (or near-perpendicular) to the incident radiation (as indicated by a dosimeter angle of 0° in Fig. 4, far left). However, the dosimeter could also be positioned such that the active volume is “edge-on” to the x-ray tube, at a 90° position (Fig. 4, far right; the top of the head would be an example of such an orientation), or at some angle in between. Six OSLD were irradiated at each of the following angles: 0° , 45° , and 90° , and the signal from each OSLD was normalized to an ion chamber response under identical scan conditions (the ion chamber was not tilted). Measurements were conducted both in air and inside the phantom (16 cm CTDI phantom) to determine the impact of different scatter environments.

2.D. Beam quality dependence (k_Q)

An energy correction factor (k_Q) is necessary in cases where the beam quality used to irradiate the nanoDots differs from the beam quality used to calibrate the dosimeter and establish the system calibration factor (N_D). In this set of experiments, a 120 kV CT scan with the dosimeter placed in air at isocenter was used as the reference/calibration condition; any other scan condition or measurement location would involve a slightly different x-ray spectrum and therefore require some energy correction. Fundamentally, this correction factor

FIG. 3. Experimental setup for determining effect of rotating beam irradiation compared to a static beam.

FIG. 4. Experimental setup for determining OSLD signal dependence on angle of dosimeter relative to CT gantry.

is equal to the ratio of the dosimeter signal per delivered dose under reference conditions (for determining the system calibration) to the experimental conditions, as shown in Eq. (5), where D denotes the dose and M denotes the dosimeter signal²⁹

$$k_Q = \frac{D_{\text{exp}} M_{\text{ref}}}{M_{\text{exp}} D_{\text{ref}}} \quad (5)$$

k_Q can be determined empirically or by using cavity theory,²⁴ where the dose to the cavity is due to both electrons crossing the cavity and secondary electrons created within the cavity.^{30,31} When using the cavity theory approach, the calculated dose to the dosimeter ($D_{\text{Al}_2\text{O}_3:\text{C}}$) is the dosimeter signal, and thus, the correction factor can be defined as follows [Eq. (6)]:

$$k_Q = \frac{(D_{\text{Al}_2\text{O}_3}/D_{\text{medium}})_{\text{reference energy}}}{(D_{\text{Al}_2\text{O}_3}/D_{\text{medium}})_{\text{experimental energy}}} \quad (6)$$

To determine k_Q using cavity theory, detailed knowledge of the reference beam (used for calibration of the dosimetry system) and the experimental beam energy spectra are needed. While differences in the nominal energy of the calibration and experimental beams (e.g., 120 kV calibration vs 80 kV experimental measurements) provide obvious differences in beam quality, variations in scan and phantom parameters will also change the spectrum and therefore the dosimeter response. In particular, variations in phantom (or patient) size, scan extent (which affects the extent of scatter), as well as dosimeter position (which affects the degree of beam hardening) may all be relevant. To determine the impact of these parameters, we used a benchmarked Monte Carlo model of our GE scanner³² to simulate the photon energy spectrum as a function of kV, phantom size (body or head CTDI phantom), depth in phantom (surface, periphery, or center positions in the CTDI phantom), and scan extent. In total, 28 unique spectra were simulated, including an in-air spectrum representing the calibration conditions (120 kV scan with dosimeter placed in air at machine isocenter). The scan parameters for each of the conditions simulated are given in Table I in order of increasing mean spectral energy. For all helical scans, a pitch of unity was used.

Using the simulated spectra, the dose to medium (D_{air}) was determined per Eq. (7). In this expression, the energy fluence, Ψ_E , was calculated per discrete energy bin, and the mass energy absorption coefficients (μ_{en}/ρ) were from NIST.³³ The sum of the dose over all energies represents the total dose to air,

$$D_{\text{air}} = \sum_E \Psi_E \left(\frac{\mu_{\text{en}}}{\rho} \right)_{\text{air}, E} \quad (7)$$

Standard Burlin cavity theory, as described and applied previously,^{16,24} was used to calculate the corresponding dose to aluminum oxide ($D_{\text{Al}_2\text{O}_3:\text{C}}$) [Eq. (8)], where coefficients were obtained from literature sources.^{33–35} We considered the impact of both differences in stopping power ratios and mass energy absorption coefficients, which were weighted with the weighting term: $d(E)$. Calculation of $d(E)$ is described previously;^{16,24} values of $d(E)$ in Al_2O_3 ranged from 0.067 at 140 keV to less than 0.001 at 10 keV. That is, differences in mass energy absorption coefficients (photon interactions) highly dominated (~98%) the energy response of the OSLD. Application of cavity theory at low energies for this detector is confounded by the fact that the electrons do not penetrate the layer of ABS plastic that encases the detector. Therefore, the relevant stopping power ratio is between Al_2O_3 and ABS plastic rather than air. However, calculating dose contributions to different media (air for photon interactions and ABS plastic for electron interactions) does not provide a physically meaningful answer. Therefore, given that the electron contribution, $d(E)$, is only around 2%, and given that air and plastic do not have wildly different stopping powers, we related all Al_2O_3 signals to air,

$$D_{\text{Al}_2\text{O}_3:\text{C}} = D_{\text{air}} \cdot \sum_E \left[d(E) \frac{\left(\frac{\bar{S}}{\rho} \right)_{\text{Al}_2\text{O}_3:\text{C}}(E)}{\left(\frac{\bar{S}}{\rho} \right)_{\text{air}}(E)} + (1 - d(E)) \frac{\left(\frac{\mu_{\text{en}}}{\rho} \right)_{\text{Al}_2\text{O}_3:\text{C}}(E)}{\left(\frac{\mu_{\text{en}}}{\rho} \right)_{\text{air}}(E)} \right] \quad (8)$$

The doses to both air and aluminum oxide were determined for both experimental spectra and the calibration spectra such

TABLE I. Scan parameters, CTDI phantom, position in phantom, and mean spectral energy for conditions simulated using a benchmarked MCNP model of the GE VCT scanner. Center and peripheral positions in phantom correspond to center and peripheral (1 cm depth) chamber positions in the CTDI phantoms.

kV	CTDI phantom size (cm)	Position in phantom	Scan type	Scan extent (mm)	Mean spectral energy (keV)
120	In-air	N/A	Axial	40	59.9
80	16	Center	Helical	150	43.8
80	16	Center	Axial	40	45.4
80	16	Peripheral	Helical	150	45.9
80	16	Peripheral	Axial	40	46.8
80	16	Surface	Helical	150	47.9
80	16	Surface	Axial	40	48.6
120	32	Center	Helical	150	51.7
120	32	Center	Axial	40	54.7
140	32	Center	Helical	150	55.0
120	16	Center	Axial	40	55.3
140	16	Center	Helical	150	55.4
120	32	Peripheral	Helical	150	57.8
120	16	Peripheral	Axial	40	57.9
140	16	Center	Axial	40	58.1
140	32	Center	Axial	40	58.7
120	32	Periphery	Axial	40	59.5
140	16	Peripheral	Helical	150	59.8
120	32	Surface	Helical	150	60.7
120	16	Surface	Axial	40	60.8
140	16	Peripheral	Axial	40	61.5
120	32	Surface	Axial	40	62.2
140	32	Peripheral	Helical	150	62.7
140	16	Surface	Helical	150	63.2
140	16	Surface	Axial	40	64.9
140	32	Peripheral	Axial	40	65.0
140	32	Surface	Helical	150	66.4
140	32	Surface	Axial	40	68.3

that a theoretical beam quality correction factor (k_Q) was determined for each experimental spectra simulated.

To verify the calculated beam quality correction factors, we also measured these correction factors for a subset of the simulated experimental conditions. All measurements were conducted using standard acrylic CTDI phantoms using both nanoDots and a farmer-type ion chamber. The dose to the ion chamber was determined using Eq. (1).

The measured beam quality correction factor was the ratio of the signal of the OSLD to the measured ion chamber dose for the reference or calibration condition (120 kV in air) relative to the experimental condition,

$$k_Q = \frac{(D_{\text{OSLD}}/D_{\text{ion chamber}})_{\text{reference energy}}}{(D_{\text{OSLD}}/D_{\text{ion chamber}})_{\text{experimental energy}}}. \quad (9)$$

3. RESULTS

3.A. Basic characterization

3.A.1. Signal depletion (k_d)

On average, the nanoDot signal was depleted by 1.7% per readout when the strong LED (low dose) beam was used for

a 7 s readout. Figure 5 displays the average depletion per reading for the five nanoDots, with a negative exponential curve fit to the data ($R^2 = 0.997$). The error bars represent the standard deviation of the signal depletion for the five dosimeters included in the study.

FIG. 5. Signal depletion over sequential readings of OSLD nanoDots.

Signal depletion is specific to the strength of the LED used and is therefore reader-specific. It will also differ if the reading time is not 7 s. It should therefore be characterized for any new reader for the appropriate dose scale (strong vs weak LED beam) before performing CT dosimetry.

3.A.2. Reproducibility

The average coefficient of variation (CoV) observed between the 20 dosimeters was 1.5% for the ~5 mGy irradiations. With a larger signal, this variability decreased to 0.7% for CT irradiations (~50 mGy) and 0.8% for ^{60}Co irradiations (~90 mGy) (Table II). These values are comparable to the average CoV for dosimeters irradiated to a high dose (1000 mGy) using ^{60}Co as part of IROC Houston's remote verification program (representing a fully commissioned batch of over 30 000 OSLD nanoDots)³⁶ which has an interdot variability of 0.8% (Table II). The maximum CoV of the three successive readings of any of the 20 dots was also reported (the maximum intradot variability). This value was consistent across all dose levels.

3.A.3. Signal fading (k_F)

The investigation of the fading of the dosimeter signal indicated a stable signal 30 min to 2 weeks after irradiation. Figure 6 shows the dosimeter response as a function of time since irradiation.

Consistent with other studies in radiotherapy environments,^{10,26} there was no significant trend seen for fading. Based on a regression fit of the data, fading over this time period was less than 3% (2-sigma).

3.A.4. Dose linearity (k_L)

The raw count reading per dose is shown in Fig. 7(a) for both the strong and weak beams of the OSLD reader. Linear trend lines were added over the low-dose component of the data (up to 600 mGy for the strong beam and 300 mGy for the weak beam). Error bars are included based on the average standard deviation of the readings but do not include the uncertainty in the dose delivery of the CT scanner. A plot of the ratio of the measured signal to the linear fit of the data [from 7(a)] is shown in Fig. 7(b) for the average of the

Fig. 6. Signal fading as a function of elapsed time since irradiation.

two measurements at each dose. For doses above 300 mGy, the strong beam increasingly saturates the reader, and the signal is seen to decrease by as much as 10% compared to a linear prediction. This effect is clearly important, but relates to saturation of the reader system, and is not related to intrinsic supralinearity of the detector. The weak beam shows a linear relationship up to at least 600 mGy, although supralinearity is apparent when the high dose data are included ($p < 0.001$). At doses below this, results were consistent with a linear response within $\pm 2\%$.

3.B. Dependence on irradiation geometry (k_G)

The relative response of the dosimeter using a rotational irradiation versus a static irradiation (Fig. 3), in air, inside of the phantom, and on the phantom surface are given in Table III.

The results of these measurements indicate that dosimeters irradiated using a rotating CT beam have a slightly lower signal than those irradiated with a static beam (perpendicular to the dosimeter), consistent with what would be expected from the literature.^{17,23} These results indicate that dosimeters which are irradiated in a rotational environment but have a calibration factor defined using a static beam should be corrected to compensate for lost signal. This effect is independent of kV, and an irradiation geometry correction factor (k_G) of 1.03 (correcting for the average under-response of the dosimeter in rotational irradiations) is recommended if the calibration dosimeters are irradiated in a planar fashion instead of a rotational CT environment. The uncertainty in this correction factor is estimated from the uncertainty in the underlying measurements to be 2% (2-sigma).

3.C. Angular dependence (k_θ)

The variation of signal based on the orientation of the dosimeter relative to the CT bore (Fig. 4) was described as angular dependence (all these tests were performed under rotating conditions). The normalized OSLD signal for dosimeters placed at non-normal angles is shown in Table IV

TABLE II. Inter- and intradot coefficient of variation (CoV) describing the variability for low-dose irradiations using both CT and ^{60}Co and for high-dose irradiations using ^{60}Co .

	CT irradiations	CT irradiations	^{60}Co irradiations	^{60}Co irradiations
Beam energy	120 kV	120 kV	1.25 MeV	1.25 MeV
Dose delivered (mGy)	5	50	90	1000
Interdot CoV (%)	1.5	0.7	0.8	0.8
Maximum intradot CoV (%)	1.7	1.7	2.0	2.0

FIG. 7. Dose-linearity response for the strong beam (low dose) and the weak beam (high dose) modes. (a) OSLD reading as a function of delivered dose, including a linear fit based on the low doses. (b) Regression plot showing the ratio of measured signal versus the signal predicted from the linear fit in (a) as a function of delivered dose.

for the four conditions considered. k_Q is the reciprocal of the relative responses in Table IV.

For dosimeters placed in full scatter conditions (in the center chamber of the phantom), there was no significant angular dependence of the OSLD signal at any angle. However, when dosimeters are in air (no scatter medium present), there is a marked decrease in signal when the dosimeter is placed at a 90° angle to the CT bore, i.e., is edge-on to the rotating tube. This behavior was previously observed by Lavoie *et al.*¹⁸ We also found this effect to be more pronounced for lower energies, showing a 16% loss in signal relative to the signal for a normally (0°) irradiated dosimeter at 80 kV, and an 11% loss in signal at 120 kV. For angles less than 90° , such as the 45° dosimeter angle, the angular dependence is within the measurement uncertainty, estimated to be 2% (2-sigma).

3.D. Beam quality dependence (k_Q)

The photon energy spectra for 28 different scan configurations (described in Table I) were calculated, including different kV, phantom sizes, measurement positions, and for both short and long scan extents, providing a range of different beam qualities. A subset of 120 kV spectra is shown in Fig. 8.

Differences in spectra resulted in differences in the response of the nanoDot calculated using Burlin cavity theory. The range of k_Q relative to the reference spectra was 0.80–1.15 for all spectra simulated (Fig. 9).

The beam quality correction factor (k_Q) determined using Burlin cavity theory was compared to that determined using direct measurement for 11 scan conditions (shown in Table V).

TABLE III. Ratio of full rotation OSLD signal to static beam OSLD signal (see Fig. 3) under various conditions for three nominal CT energies.

kV	Measurement condition		
	In air	On phantom surface	In the center of phantom
80	0.98	0.97	0.99
120	0.98	0.99	0.97
140	0.98	0.95	0.97

The average magnitude of percentage difference between the measured and calculated values of k_Q was 4%.

Over the range of scan parameters tested, the variation in beam quality relative to the calibration condition led to a variation in OSLD response ranging from -20% (80 kV, center of a 16 cm phantom) to $+15\%$ (140 kV, surface of a 32 cm phantom). Nominal kV had the largest influence on k_Q , followed by the position in the phantom (center position vs surface position). Changing the phantom size alone (kV, scan extent, and dosimeter position remaining the same) impacted the magnitude of the correction factor by only 1%–5%. Differences in scan extent (single axial rotation vs 15 cm scan length) also had only a small effect (1%–4%) on the correction factor. The two-sigma uncertainty in k_Q was estimated to be less than 5% across the conditions investigated.

4. DISCUSSION

NanoDot OSLDs were characterized in a CT environment. In some aspects, the dosimeter behaved differently than in high-energy and high-dose applications. In other aspects, it behaved very similarly. The reproducibility of this dosimeter was $\sim 0.7\%$ at doses as low as 50 mGy and deteriorated slightly to 1.5% at 5 mGy. At such low doses, the dot's bleaching history (i.e., any residual signal) and the specific reader's noise level will be relevant, and the specific low-dose variability will therefore be system dependent.

Depletion was found to be substantial enough to warrant consideration, although this effect would be mitigated with use of the more common 1 s read-time.

The signal was found to be stable for the time interval examined in this study and no correction factor for fading would be necessary during the time period of 30 min to 2 weeks after irradiation. However, as some fading does, in reality, occur; application of a correction may be necessary if nanoDots are intended to be used as a long-term dosimetric record.

In high-energy beams, Jursinic and Reft observed a linear response up to 2000 mGy.^{10,26} In the current study, nonlinearity effects (although small) were apparent around 700 mGy, which is a much lower dose than that reported

TABLE IV. Relative OSLD signal for dosimeters oriented at 0°, 45°, and 90° relative to the signal at 0° (see Fig. 4) for two nominal energies and two irradiation conditions (in air and in the center of the 16 cm diameter CTDI phantom).

Dosimeter angle (deg)	80 kV		120 kV	
	In air	In the center of phantom	In air	In the center of phantom
0	1.00	1.00	1.00	1.00
45	0.98	1.02	0.98	0.98
90	0.84	1.00	0.89	1.00

for high-energy beams. However, these findings are actually consistent. OSLD over-responds in a CT setting compared to a megavoltage by a factor of more than 3.²⁶ Therefore, 700 mGy of kV x-rays will generate roughly the same signal as 2000 mGy of MV x-rays. These findings warrant attention when using this detector in a CT environment. At low doses (<300 mGy), either reading intensity will provide a highly linear dose response. At doses above 300 mGy, the high intensity scale can start inducing substantial saturation of the reader. The low-intensity mode does not saturate at clinical doses, but at doses above 600 mGy, nonlinearity on the order of a few percent manifests. This may warrant application of a k_L correction if high doses are encountered and very high precision is desired.

There are conflicting results on the matter of angular or orientation issues in recent literature. In CT dosimetry, both irradiation geometry (k_G) and angular dependence (k_θ) must be considered and should be considered separately. First, differences in the irradiation geometry (k_G) of experimental dots (exposed to a rotating beam) versus calibration dots (if they are exposed to a static beam) showed a 3% effect regardless of the amount of scatter or kV. Thus, a correction factor of 1.03 is appropriate if the calibration dosimeters are irradiated in a static (normal) geometry while the experimental dosimeters are irradiated with a rotating geometry. If all nanoDots are irradiated under the same condition (e.g., both

static or both rotating), no correction is needed. Second, the angular dependence (k_θ) of the dosimeter, relating the actual physical angle of the dosimeter relative to the CT beam, may need consideration. For measurements performed in air, a loss in signal was noted for dosimeters positioned 90° to the CT beam, or edge-on, and was more pronounced for lower energies. Dosimeter angles of 45° or less had minimal signal loss that was within the measurement uncertainty. Similarly, the angular dependence was not greater than measurement uncertainty for measurements at any angle performed in full scatter conditions (in-phantom). Clinical measurements would most likely be performed with the dosimeter placed on the phantom or patient surface. The surface condition represents an “intermediate” scatter condition between that seen in air and in full scatter conditions. Unfortunately, it is extremely difficult to duplicate surface scatter conditions while varying the dosimeter angle, i.e., isolating for the effect of angular dependence at the surface. Nevertheless, clinical recommendations can be made based on the results presented in Table IV. An angular dependence correction (k_θ) is not necessary for dosimeter angles less than 45°, for measurements made in air, inside of the phantom, or on the surface. Dosimeters that are placed edge-on (90°) on the surface of a phantom will likely under-respond. While a fully edge-on measurement on the surface is unlikely, it is prudent to add a correction factor to compensate for signal

FIG. 8. Monte Carlo simulated spectra for a 120 kV beam, 32 cm CTDI phantom, and two scan extents: a single axial rotation using a 40 mm beam width and a 15 cm helical scan extent (also using a 40 mm beam width).

FIG. 9. Value of beam quality correction factor (k_Q) for all simulated scan conditions as a function of mean spectral energy. See Table I for energy spectra and scan conditions. The reference location had a mean spectral energy of 59.9 keV (see Table I).

TABLE V. Comparison of measured and calculated beam quality correction factors for a range of clinical CT scanning parameters. Peripheral position denotes chamber located at 1 cm depth from phantom surface. Mean spectral energy is defined at the dosimeter position.

CTDI phantom size (cm)	Position in phantom	kV	Scan extent (mm)	Mean spectral energy (keV)	Calculated k_Q	Measured k_Q	Difference (%)
16	Center	80	40	45.4	0.82	0.85	3.2
16	Surface	80	150	47.9	0.85	0.86	0.8
32	Center	120	150	51.7	0.88	0.92	4.0
32	Center	120	40	54.7	0.91	0.95	4.0
16	Center	120	40	55.3	0.93	0.97	4.3
16	Center	140	40	58.1	0.97	1.03	5.9
32	Center	140	40	58.7	0.96	1.01	4.9
32	Periphery	120	40	59.5	1.00	1.00	0.0
32	Surface	120	150	60.7	1.02	0.95	6.9
16	Surface	140	150	63.2	1.06	1.02	3.9
16	Surface	140	40	64.9	1.09	1.02	6.8

loss and also recognize that such a measurement will have larger uncertainty.

Variations in the photon energy spectra between the calibration condition and the measurement conditions may substantially impact the response. For calibrations performed using a 120 kV CT beam in air, the spectral variations manifested as a difference in dosimeter response between +20% and -15%. Correction factors based on kV, phantom conditions, and scan length were generated using two different methods, which agreed with each other reasonably well. kV, which has previously been identified as a relevant factor,^{18,20} had the greatest impact on k_Q , while the location in the phantom (scatter conditions) could also substantially impact the dosimeter response.

5. CONCLUSION

The majority of possible correction factors for accurate OSLD were small (<5%). The dosimeter response was found to be independent of dose nonlinearity effects for the range of doses expected in CT dosimetry, and no fading correction is necessary for dosimeters' readout between 30 min and 2 weeks following irradiation. A reader-specific signal depletion correction factor should be evaluated because the high-powered light required to read out CT doses reduces dosimeter signal. Furthermore, saturation of the reader can occur and thereby underestimate the delivered dose. Two correction factors were identified that could necessitate substantial corrections: angular dependence and beam quality dependence could impact the signal by up to 16% and 20%, respectively. Corrections based on specific measurement and calibration conditions are likely necessary for these parameters when using this dosimeter in a CT environment.

ACKNOWLEDGMENTS

This work was funded in part by Public Health Service Grant No. CA 10953 awarded by the National Cancer Institute, support from Landauer, Inc., a grant from the American Legion Auxiliary (Scarboro), the PEO Scholar Award (Scarboro), and

Cancer Center Support Grant No. P30 CA016672 (Stingo). The authors report no conflicts of interest in conducting the research.

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